(19) Europäisches Patentamt
European Patent Office
Office européen des brevets

(11) EP 0 747 004 A2

(12) EUROPEAN PATENT APPLICATION

(43) Date of publication:
11.12.1996 Bulletin 1996/50

(51) Int. Cl.[6]: **A61B 5/026**, G01R 33/56

(21) Application number: **96202027.7**

(22) Date of filing: **18.09.1992**

(84) Designated Contracting States:
DE FR NL

(30) Priority: **14.11.1991 US 791855**

(62) Application number of the earlier application in accordance with Art. 76 EPC: **92308551.8**

(71) Applicant: **PICKER INTERNATIONAL, INC.**
Cleveland, Ohio 44143 (US)

(72) Inventors:
- **Woods, Christopher H.**
Chesterland, Ohio 44120 (US)
- **Apicella, Anthony**
Willoughby, Ohio 44092 (US)
- **Nessaiver, Moriel S.**
Cleveland Heights, Ohio 44118 (US)

(74) Representative: **McGowan, Nigel George**
The General Electric Company plc
GEC Patent Department
Waterhouse Lane
Chelmsford, Essex CM1 2QX (GB)

Remarks:
This application was filed on 16 - 07 - 1996 as a divisional application to the application mentioned under INID code 62.

(54) Method of measuring ventricular volumes

(57) A first image and a second image of a patient's heart region are taken at small time displaced intervals. The first and second images are subtracted to generate a difference image which is indicative of the tissue which has moved during the short time interval, i.e. the boundary of the ventricles. Voxels from regions outside the boundary are adjusted to remove lung tissue, and ventricle boundary or edge voxels and analyzed to generate a non-blood tissue histogram (62). Voxels within the boundary are analyzed to generate a blood tissue histogram (60). The histograms (60, 62) are fitted to smooth curves (68, 76) which represent the probability distribution of whether each voxel value represents blood or no-blood tissue. Contiguous voxels within the boundary are counted and adjusted for voxel size to create an indication of left and right ventricle volume. In the preferred embodiment, the ventricle volume is determined by summing the confidence value that each voxel within the boundary represents blood.

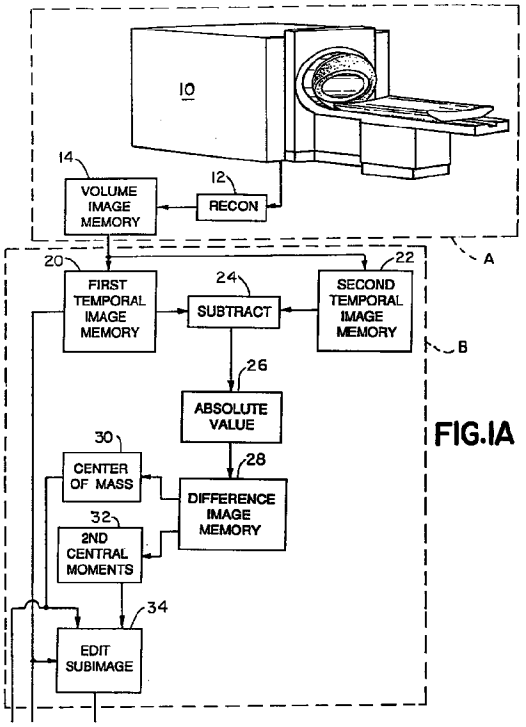

FIG.1A

EP 0 747 004 A2

(Cont. next page)

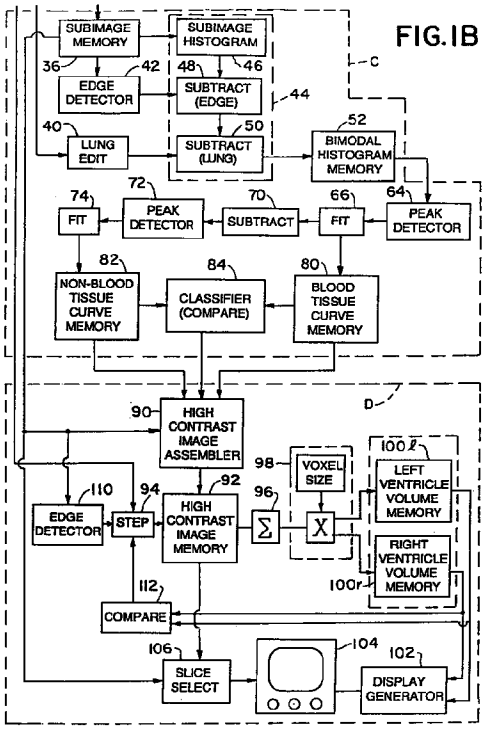
SUBIMAGE MEMORY
SUBIMAGE HISTOGRAM
36
42
48
46
EDGE DETECTOR
SUBTRACT (EDGE)
44
C
40
LUNG EDIT
SUBTRACT (LUNG)
50
52
BIMODAL HISTOGRAM MEMORY
74
72
70
66
64
FIT
PEAK DETECTOR
SUBTRACT
FIT
PEAK DETECTOR
82
84
80
NON-BLOOD TISSUE CURVE MEMORY
CLASSIFIER (COMPARE)
BLOOD TISSUE CURVE MEMORY
D
90
HIGH CONTRAST IMAGE ASSEMBLER
98
VOXEL SIZE
100l
LEFT VENTRICLE VOLUME MEMORY
110
94
92
96
EDGE DETECTOR
STEP
HIGH CONTRAST IMAGE MEMORY
Σ
X
RIGHT VENTRICLE VOLUME MEMORY
100r
112
COMPARE
106
104
102
SLICE SELECT
DISPLAY GENERATOR

FIG.1B

Description

The present invention relates to medical measurement techniques. It finds particular application in conjunction with measuring the ventricular volumes in a patient's heart from magnetic resonance image data and will be described with particular reference thereto.

This application is divided out of European Patent Application No.92308551.8.

Heretofore, ventricular volumes have been determined by first generating a volumetric image representation of a rectangular portion of the patient's body that includes the heart. Commonly, the volume image representation included a plurality of parallel planar or slice images which were coordinated with the cardiac cycle such that each slice or plane was taken at or near the same cardiac phase. A trained radiologist or technician viewed each slice and marked the edges of the ventricles. Once the boundary of the ventricle regions was marked, the volume could be readily calculated. For example, the number of voxels in the volume within the boundary or the number of voxels of each slice within the boundary could be counted. From the dimensions of the voxel or the dimensions of the voxels and the interslice spacing, the volume was readily calculated. Not only did this technique require a large amount of time, but the results were not reproduceable. Each radiologist or other trained expert tended to define the edges of the ventricles differently.

Various automatic methods have been proposed for determining the ventricular volumes without human assistance. Because the blood and the tissue have different gray scale or intensity, one could use this difference to determine the boundary. Boundaries or edges of the magnetic resonance image are characterized by high frequency components. Noise, motion, and other artifacts are present at all frequencies but most significant in the high frequency components. This noise created uncertainty in the location of the interface. In one approach, a contour following algorithm was applied to the image to make a first approximation of the boundary of the ventricle in each slice. Density profiles on either side of the proposed border were examined and the contour following algorithm was reapplied. This procedure was repeated iteratively until a stable threshold was obtained. This iterative approach was not only time consuming and computationally expensive, but tended to have inaccuracies arising from the inclusion of surrounding tissue other than ventricles in the process.

Another drawback to this method is that it assumed that the intensity profile passing through the heart/blood interface was sigmoidal, with the optimal edge position existing at voxels having the largest gradient. This assumption is unsupported. When the intensity profile is not sigmoidal, there are errors in the estimate of the cardiac volume.

In a second technique, a threshold was used to create a binary image from the volume image. That is, the intensity or gray scale of each voxel was examined and compared with a threshold. Based on this comparison, each pixel or voxel was classified as blood, hence an interior region of the ventricle, or non-blood. Identifying the threshold value commonly required an initial human guess. The guess was iteratively adjusted in repetitions of the binary image forming process until appropriate results were achieved. This technique was again slow and computationally burdensome.

Another technique used a priori probabilities for the expected tissue classes within the volume to produce an initial segmentation of the image, i.e. define the ventricle boundaries. These probabilities were updated in accordance with the initial image and the segmentation process repeated with the updated probabilities. This process was iteratively repeated until the segmentation reached a steady state. This technique required some manual segmentation to create a training set and to determine the initial probabilities. Moreover, modifying the decision criteria, i.e. the probabilities, based on results tended to perpetuate inaccuracies.

In another approach, each voxel was classified based on its gray scale level, gradient, and location. The voxel classifications were iteratively updated in order to make the relative clarifications more consistent using the Dempster-Schafer theory of knowledge. This technique again required a priori information and was computationally excessive. Moreover, because the iterations were based on prior iterations, errors were compounded.

The present invention provides a new and improved volume measurement technique which alleviates the above-referenced problems and others.

According to a first aspect of the invention, there is provided a method of measuring ventricular volumes comprising: non-invasively examining a region of a patient which includes at least one ventricle to generate values of voxels of a ventricle image representation; determining which voxels of the ventricle image representation are indicative of a peripheral boundary of the ventricle; for at least a portion of the voxels within the voxels which define the ventricular boundary, determining a frequency of occurrence for each inside voxel value; in accordance with the frequency of occurrence of each inside voxel, determining a probability that each voxel value corresponds to blood tissue; counting inside voxels and weighting each counted voxel in accordance with its determined probability to determine a sum which is proportional to the ventricular volume.

According to the second aspect of the invention, there is provided an apparatus for measuring ventricular volumes comprising: non-invasive examination means for non-invasively examining a region of a patient which includes at least one ventricle to generate values of voxels of a ventricle image representation; means for determining which voxels of the ventricle image representation are indicative of a peripheral boundary of the ventricle; histogram generating means for determining a frequency of occurrence of values of voxels surrounded by the ventricle boundary; means for deter-

mining from the frequency of occurrence a probability that each voxel value corresponds to blood tissue; counting means for counting voxels surrounded by the ventricle boundary and weighting each counted voxel in accordance with its determined probability to determine a sum which is proportional to the ventricular volume.

One advantage of the present invention resides in its fast and simple, yet accurate segmentation.

Another advantage of the present invention resides in the improved accuracy. Eliminating lung and uncertain edge values before determining the criteria for classifying a voxel as blood and non-blood eliminates a major source of error in the resultant images.

Another advantage of the present invention is that it produces accurate results even with noisy images. Making voxel classification decisions based on a histogram as opposed to individual voxel properties improves overall reliability.

Other advantages of the present invention reside in its improved processing speed and reduced complexity. Iterative data analysis is eliminated, as are training sets and the inputting of a priori information or a knowledge base.

One embodiment of a method and apparatus according to the invention will now be described with reference to the accompanying drawings in which:

Figures 1A and 1B illustrate an embodiment of the present invention;

Figure 2 illustrates exemplary blood and non-blood tissue probability curves.

With reference to Figure 1, a magnetic resonance imaging apparatus A examines a selected region of a subject and generates a volume image representation thereof. The volume image representation of the heart and surrounding areas, is a product of a black blood cine sequence in which the volume is defined by a plurality of parallel slices. A motion detection means B locates the region of the ventricles. A segmentation means C parametrically determines intensity ranges of voxels of the image representation in a blood intensity range and voxels in a non-blood tissue intensity range. A region growing means **D** collects or counts contiguous image voxels which have the intensity of blood volume of the image representation.

The magnetic resonance data acquisition means **A** includes a conventional magnetic resonance scanner **10**. The magnetic resonance scanner includes the appropriate coils and supporting electronics for generating a substantially uniform main magnetic field through an imaging field. Radio frequency coils and supporting electronics selectively introduce radio frequency pulses into the examination region to induce resonance of selected bipoles and to manipulate the induced resonance. Gradient field coils and supporting electronics apply gradient field pulses across the examination region to provide phase and frequency encoding in the excited resonance field and for defining slices or volumes.

Magnetic resonance echoes or other diagnostic data is collected and reconstructed by a reconstruction means **12** into the volume image representation which is stored in an appropriate volume image memory **14**. The reconstruction means **12** reconstructs a series of two dimensional slices which taken together define the volume. Alternately, a three dimensional reconstruction algorithm may be used. In the preferred black blood cine sequence, blood tissue appears at one intensity extreme, the intensity extreme which is commonly displayed as black on a black and white monitor. Non-blood tissue contributions to the image tend to be near the other intensity extreme, the intensity extreme which is generally displayed as white. Preferably, the black blood cine technique is used to generate a plurality of planes collected at the same point of the cardiac cycle. Obtaining high temporal resolution is attained by using cardiac gating. That is, the patient's cardiac cycle is monitored and the MRI scan commences a preselected duration after the R-wave or other characteristic portion of the cardiac cycle. Multiple measurements (up to 64) are acquired at a fixed interval (typically 10-25 msec). The preferred black blood cine sequence results in temporal resolution of 20 msec. which allows for 40 images covering 80% of a typical 1000 msec. R-R interval. Optionally, CT or other data acquisition means may be utilized.

The motion detecting means **B** causes two temporally adjacent images to be generated. That is, the black blood cine sequence is used to generate the same spatial volume, but offset a very short time interval. The second MRI scan is triggered a duration after the R-wave which is slightly longer (or shorter) than the duration after the R-wave at which the first image was generated. The motion means **B** includes a first memory means **20** for storing a first temporal image and a second temporal memory means **22** for storing a second temporal image. The first and second memory means are preferably portions of a large RAM or disc memory. A subtracting means **24** subtracts corresponding voxels of the two images generating a temporal difference image. Because the first and second temporally displaced images are taken very close together in time, they are essentially identical, except in areas of movement. The movement, of course, is primarily in the ventricles. Thus, the difference image is essentially blank, except for a dark line around the ventricles, which dark line has a width which substantially corresponds to the amount of cardiac movement in the time between the first and second images. An absolute value means **26** compares the absolute value of each voxel of the difference image with a preselected threshold value selected to differentiate between the differential ventricle movement surface or band line and stray differences. Small values are set to zero and other values are set to one. Optionally, an additional algorithm may be incorporated for discarding voxels which are remote from the contiguous surface of voxels surrounding the ventricles. The

absolute value of the difference image is stored in a difference image memory means **28**.

The center of mass of the temporal difference image falls at or near the boundary between the left and right ventricles in a short axis view of a normal human heart. The second central moments of the difference image correlate directly to how large a subimage is necessary in order to contain just the heart. A center of mass means **30** determines the center of mass or geometric center from the temporal difference image. A second moment means **32** calculates the second moments of the difference image. An editing means **34** edits the original volume image, slice by slice, from the first temporal image memory means **20** to produce a square cross section subimage which contains substantially just the heart. That is, the subimage is centered at the center of mass of the difference image and has side lengths equal to the root of the second central moments of the difference image at the largest cross section of the heart. Optionally, the center and image size information can be used to reformat the MRI sequence to limit the examination region to the subimage region.

The segmentation means **C**, in the preferred embodiment, segments or identifies three types of tissue - blood, lung, and other types of non-blood tissue. Prior to the determination of probabilities, a lung editing means **40** examines the voxels of the original volume image in memory **20** to locate the lung region. Prior to computation of the histogram, all voxels in the lung region are excluded from the computation. Lung location is determined by region growing to collect background voxels over the entire images. The largest regions are along the left and right sides of the patient and the next largest region is the cavity containing the lung. An edge detection means **42** detects edge voxels corresponding to edge regions in the selected subimage in the subimage memory means **36**. An edge image is computed by passing a sobel operator over the subimage and thresholding the results. The resulting histogram now contains primarily information related to blood and tissue only. This gives a strong binomial flavor and assists in clustering the histogram to determine probabilities for blood and tissue.

A subimage histogram means **44** generates a histogram, i.e. intensity versus number of voxels, for voxels in the blood region and in the non-blood, non-lung, non-edge region. More specifically, a subimage histograms means **46** determines the intensity corresponding to each voxel of the subimage stored the subimage memory **36**. An edge voxel subtracting means **48** subtracts or deletes the contributions from the edge voxels as determined by the edge detection means **42**. In this manner, voxels which contain part blood and part non-blood tissue are excluded from being part of the basis of the projected probabilities. Analogously, a lung voxel subtracting means **50** subtracts or deletes the contribution of the voxels or voxels identified by the lung region identifying means **40** as being lung tissue. A bimodal histogram memory means **52** stores the histogram of voxel intensity versus number of occurrences of the blood and of the non-blood, non-lung, non-edge voxels.

With continuing reference to FIGURE 1 and further reference to FIGURE 2, the histogram defines a curve with two peaks, one peak **60** corresponding to the average blood tissue intensity and the other peak **62** centered around the average non-blood (non-lung, non-edge) tissue intensity. The low or black intensity cluster of voxels represents blood and the high or light intensity cluster represents non-blood tissue. A first peak detector **64** detects one of the peaks, e.g. the blood cluster. A first curve fitting means **66** fits a smooth curve to the blood peak of the histogram. In the preferred embodiment, the curve fitting means **66** fits the blood peak to a first Gaussian curve **68**. The intensity level occurring most frequently in the histogram is used as the initial estimate of the mean Gaussian distribution. The frequency of occurrence of the initial estimate of the amplitude and the initial estimate of the variance is estimated by tracking amplitude in the vicinity of the peak in both directions. An amplitude decrease by a factor of e, the base of naperian logarithms, marks the variance. These initial estimates of the mean and variance are entered into a non-linear least squares fit routine to estimate the Gaussian curve best representing the blood peak. A subtracting means **70** subtracts the blood tissue Gaussian curve from the bimodal histogram in the vicinity of the blood peak. A second peak detector means **72** detects the non-blood cluster peak. A second curve fitting means **74** fits a second Gaussian curve **76** to the non-blood tissue peak.

Of course, other curve fitting routines may also be used. Other appropriate curves for fitting to the histogram include:

$$lognormal\ \frac{1}{x}e^{-\ln^2 x/2} \quad (1)$$

$$Maxwell\ x^2 e^{x^2/2} \quad (2)$$

$$Erlang\ x^r e^{-x} \quad (3)$$

$$Cauchy\ 1/(1+x^2) \quad (4)$$

$$Normal\ e^{-x/2} \quad (5)$$

$$Beta\ x^b(1-x)^c \quad (6)$$

and the like. The best fit blood histogram curve is stored in a blood tissue curve memory **80** and the best fit tissue histogram curve is stored in a non-blood tissue histogram memory **82**. The blood and tissue best fit curves represent the probability that each voxel represents blood or tissue. That is, the amount of deviation between the intensity of a voxel and the peak of the blood curve or the tissue curve is indicative of the probability that the voxel represents blood or tissue, respectively. The best fit curves **68**, **76** are preferably determined based on a single slice of the volume image. The determined probability curves are valid for

other slices of the volume.

A classification means **84** compares each available voxel intensity with the probability curves **68**, **76** and classifies the voxel as either blood or non-blood tissue and the probability or certainty of the classification. That is, the apex of each of Gaussian curves **68** and **76** is set equal to one, i.e. 100% probability. For each intensity, the probability that it represents blood or non-blood tissue is indicated by the probability curve. If a given intensity has a higher probability on the blood curve than on the non-blood curve, it is classified as blood tissue. If a given intensity has a higher probability on the non-blood curve than the blood curve, it is classified as non-blood tissue.

The region growing means **D** includes a high contrast or blood image forming means **90** which determines the classification of each voxel of the subimage. More specifically, it consults the classification means **84** with the intensity of each voxel to determine whether the voxel represents blood or non-blood tissue. If the intensity represents non-blood tissue, a zero value is assigned to that voxel. If the intensity is classified as blood tissue, then the blood image forming means **90** consults the probability distribution curve memory means **80** to determine the confidence with which the corresponding intensity was classified as blood. Again, the apex of the best fit curve is classified as one or a 100% confidence value. Values further down the curve from the peak have a correspondingly lower confidence value. In the preferred embodiment, each voxel which is determined to represent blood, is assigned the corresponding confidence value or probability. Alternately, each voxel which is determined to represent blood tissue, may be given the binary value one and voxels which are determined to represent non-blood tissue may be given the value zero. As yet another alternative, those voxels which are determined to represent non-blood tissue may be given the inverse of their confidence level, i.e. the probability that the voxel does not represent non-blood tissue, i.e. represents blood tissue.

The high contrast or blood image representation is stored in a high contrast image memory means **92**.

The growing means **D** starts at a voxel which is known to be in either the left or right ventricle from the center of mass and second moment determinations made by means **30** and **32**. A stepping means **94** reads out the voxel value at the voxel which is known to be in one of the left and right ventricles. The stepping means than moves outward from this voxel, generally in concentric circles, reading the confidence value of each non-zero, contiguous voxel. The stepping means continues to step outward, but only in directions along which blood voxel was found, until all contiguous voxels which have intensities which represent blood are read out or counted. Non-contiguous voxels that are classified as blood tissue may represent other blood vessels rather than the ventricle interior and are not read out. Preferably, slices of the volume image are processed serially. The stepping means determines the initial point of the first slice based on the center of mass. Once the volume of the ventricle in that slice has been determined, the center of mass in that slice can be determined precisely. The stepping means preferably includes a means for calculating the center of mass of each slice and using the calculated center of mass as the starting point for the next contiguous slice.

A summing means **96** sums the contiguous blood voxel values which the stepping means **94** reads out of the high contrast memory **92**. This sum of the confidences is multiplied by the volume of each voxel by multiplying means **98**. This product is indicative of the volume of the ventricle, which volume is stored in a ventricle volume memory means **100**. The stepping means **94** then steps to a voxel which is predicted from the center mass to be generally centered in the other ventricle and the process repeated.

More specifically to the preferred embodiment, the ventricle volume memory means **100** includes a left ventricle volume memory means **100ℓ** and a right ventricle volume memory means **100r**. A volume display means **102** converts the stored ventricle volumes to an appropriate format for display. In the illustrated embodiment, the display means **102** converts the number into an appropriate format to be displayed on a video monitor **104**. A slice selecting means **106** enables an operator to select one or more slices of the image representation stored in the high contrast memory means **92** or the subimage memory means **36** to be displayed on the video monitor **104**.

The left and right ventricles may not appear as distinct subregions in every slice and in every cardiac phase. Thus, in slices in which the ventricles appear connected, the stepping means will tend to step through both ventricles as it moves from contiguous blood voxel to contiguous blood voxel. An edge operator routine **110** examines the initial image from memory means **20** or the subimage from memory means **36**. The high frequency components of the data which was reconstructed by the reconstructing means **12** identifies the interfaces indicative of the edge or boundary between each ventricle and the cardiac tissue. In the preferred embodiment, the edge operator routine **110** uses a sobel edge detector to determine those points lying along the edges. The edge operator based on this analysis defines the edges of each ventricle. The stepping means compares the address of each voxel which it is about to access in the high contrast memory means **92** to determine whether or not it is within or across the edge parameters determined by the edge operator routine **110**. Voxels outside of the ventricle volume indicated by the edge operator routine **110** are not accessed or read out to the summing means **96**.

A comparing means **112** compares the volumes of the left and right ventricles after the voxels of each slice are summed. By comparing the relative volumes of the left and right ventricles in each slice and by comparing the volumes of each ventricle with the preceding slice or adjacent slices, the comparing means **112** readily deter-

mines whether the stepping means has stepped between the two ventricles. When the comparing means determines that the ventricles have been combined, the comparing means enables the stepping means **94** to access the edge operator routine **110** to determine whether or not it is crossing the ventricle boundaries.

Claims

1. A method of measuring ventricular volumes comprising: non-invasively examining a region of a patient which includes at least one ventricle to generate values of voxels of a ventricle image representation; determining which voxels of the ventricle image representation are indicative of a peripheral boundary of the ventricle; for at least a portion of the voxels within the voxels which define the ventricular boundary, determining a frequency of occurrence for each inside voxel value; in accordance with the frequency of occurrence of each inside voxel, determining a probability that each voxel value corresponds to blood tissue; counting inside voxels and weighting each counted voxel in accordance with its determined probability to determine a sum which is proportional to the ventricular volume.

2. A method according to Claim 1 further including: comparing each outside voxel value with a voxel value indicative of lung tissue; eliminating outside voxels whose voxel values indicate lung tissue; for at least a portion of the remaining voxels outside of the voxels which define the ventricular boundary, determining a frequency of occurrence of each outside voxel value; determining a probability that each remaining outside value corresponds to non-lung, non-blood tissue in accordance with the determined frequency of occurrence of each outside voxel value.

3. A method according to Claim 2 wherein in the probability determining steps, the frequency of occurrence determinations are fitted to curves (68, 69), which curves are generally bell shaped and wherein the probability is determined in accordance with deviation of each, voxel value from a most frequently occurring voxel value.

4. A method according to Claim 3 wherein the counting steps includes: determining an approximate centre of the ventricle; comparing each contiguous voxel value with the probability curve (68) for blood; counting each voxel whose voxel value is indicated by the probability curve as probably being blood tissue; repeating the comparing and counting steps for voxels contiguous to each counted pixel voxel.

5. A method according to Claim 3 wherein the curve fitting technique includes: fitting frequency of determination for voxels within the ventricle boundary with a first curve; subtracting the first curve from the frequency of determination for voxels both inside and outside of the ventricle boundary to create a frequency of occurrence difference; and fitting a second curve to the frequency of occurrence difference.

6. A method according to Claims 3, 4 or 5, wherein the curve is one of a Gaussian curve, a log normal curve, a Maxwell curve, an Erlang curve, a Cauchy curve, a normal curve and a beta curve.

7. A method according to Claim 1, wherein the heart has left and right ventricles, the ventricle boundary defining boundaries of the left and right ventricles and wherein some voxels within the left and right boundaries are immediately contiguous, whereby in the counting step, there is a tendency to count voxels in both the left and right ventricle such that the sum would be indicative of a volume of the left and right ventricle and further including: determining a boundary between the left and right ventricles; during the counting step, restraining counting to opposite sides of the boundary such that the left and right ventricle volumes are determined individually.

8. A method according to Claim 1, wherein the step of determining the ventricle boundary includes: generating first and second images at a time displaced interval; subtracting the first and second images to create a difference image representation depicting the ventricle boundary and any other subregions which moved in the interval between the first and second images.

9. A method according to Claim 1 further including after determining the ventricle boundary, determining a smaller subregion which encompasses the ventricle boundary region and non-invasively examining the subregion to generate an image representation of the subregion, whereby resolution within the subregion is improved.

10. An apparatus for measuring ventricular volumes comprising: non-invasive examination means (A, B) for non-invasively examining a region of a patient which includes at least one ventricle to generate values of voxels of a ventricle image representation; means (42) for determining which voxels of the ventricle image representation are indicative of a peripheral boundary of the ventricle; histogram generating means (52) for determining a frequency of occurrence of values of voxels surrounded by the ventricle boundary; means for determining from the frequency of occurrence a probability that each voxel value corresponds to blood tissue; counting means (D) for counting voxels surrounded by the

ventricle boundary and weighting each counted voxel in accordance with its determined probability to determine a sum which is proportional to the ventricular volume.

11. An apparatus according to Claim 10 further including: means (40) for comparing values of voxels outside of the ventricle boundary with a voxel value indicative of lung tissue; means (50) for subtracting outside voxels whose voxel values indicate lung tissue; means for determining a frequency of occurrence of each outside voxel value; means for determining a probability that each remaining outside voxel value corresponds to non-lung, non-blood tissue in accordance with the determined frequency of occurrence.

12. An apparatus according to Claim 11, wherein the means for determining the ventricle boundary includes: means (24) for subtracting first and second images generated at a time displaced interval to create a difference image representation depicting the ventricle boundary.

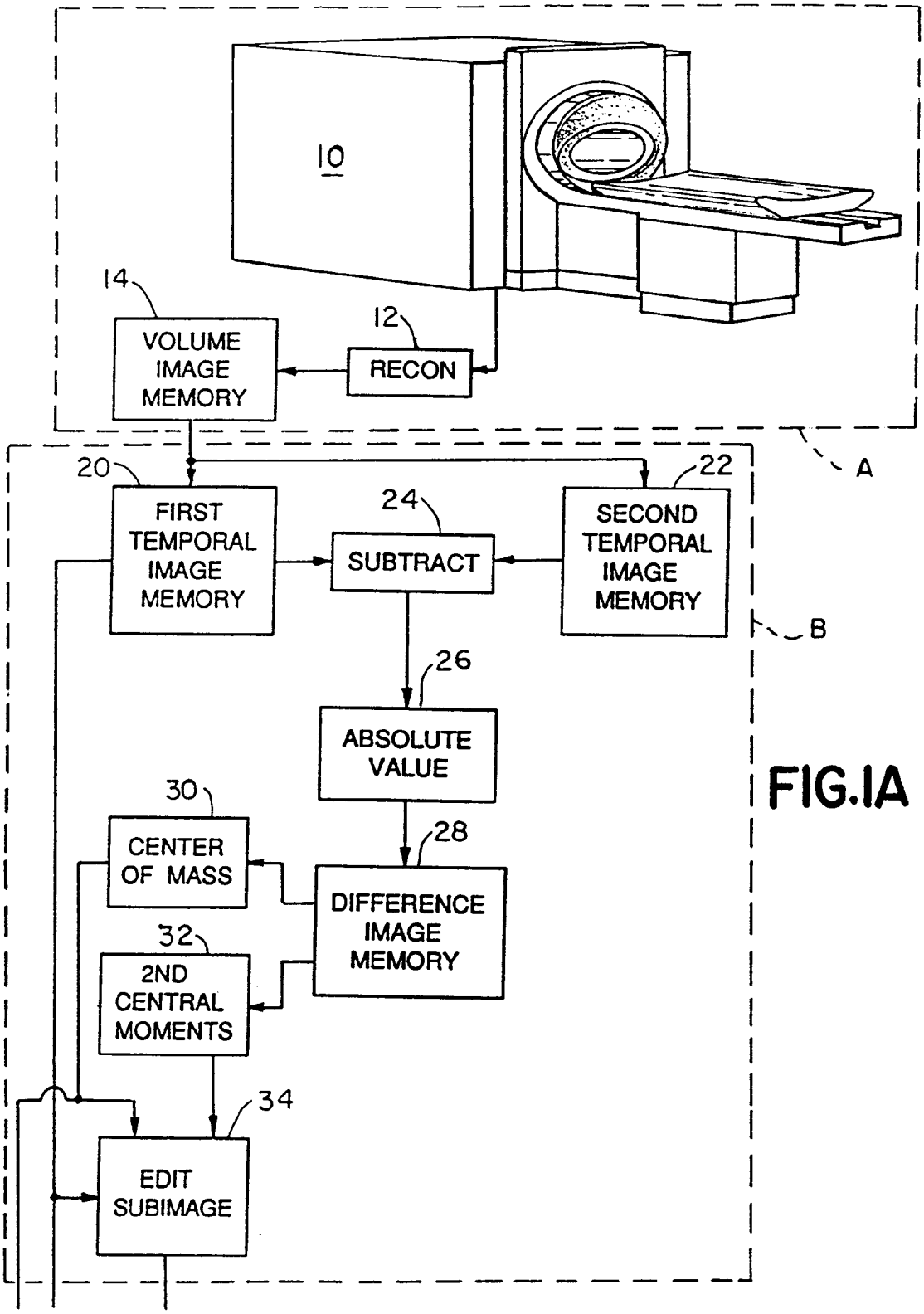
10
14
VOLUME IMAGE MEMORY
12
RECON
A
20
FIRST TEMPORAL IMAGE MEMORY
24
SUBTRACT
22
SECOND TEMPORAL IMAGE MEMORY
B
26
ABSOLUTE VALUE
30
CENTER OF MASS
28
DIFFERENCE IMAGE MEMORY
32
2ND CENTRAL MOMENTS
34
EDIT SUBIMAGE

FIG.1A

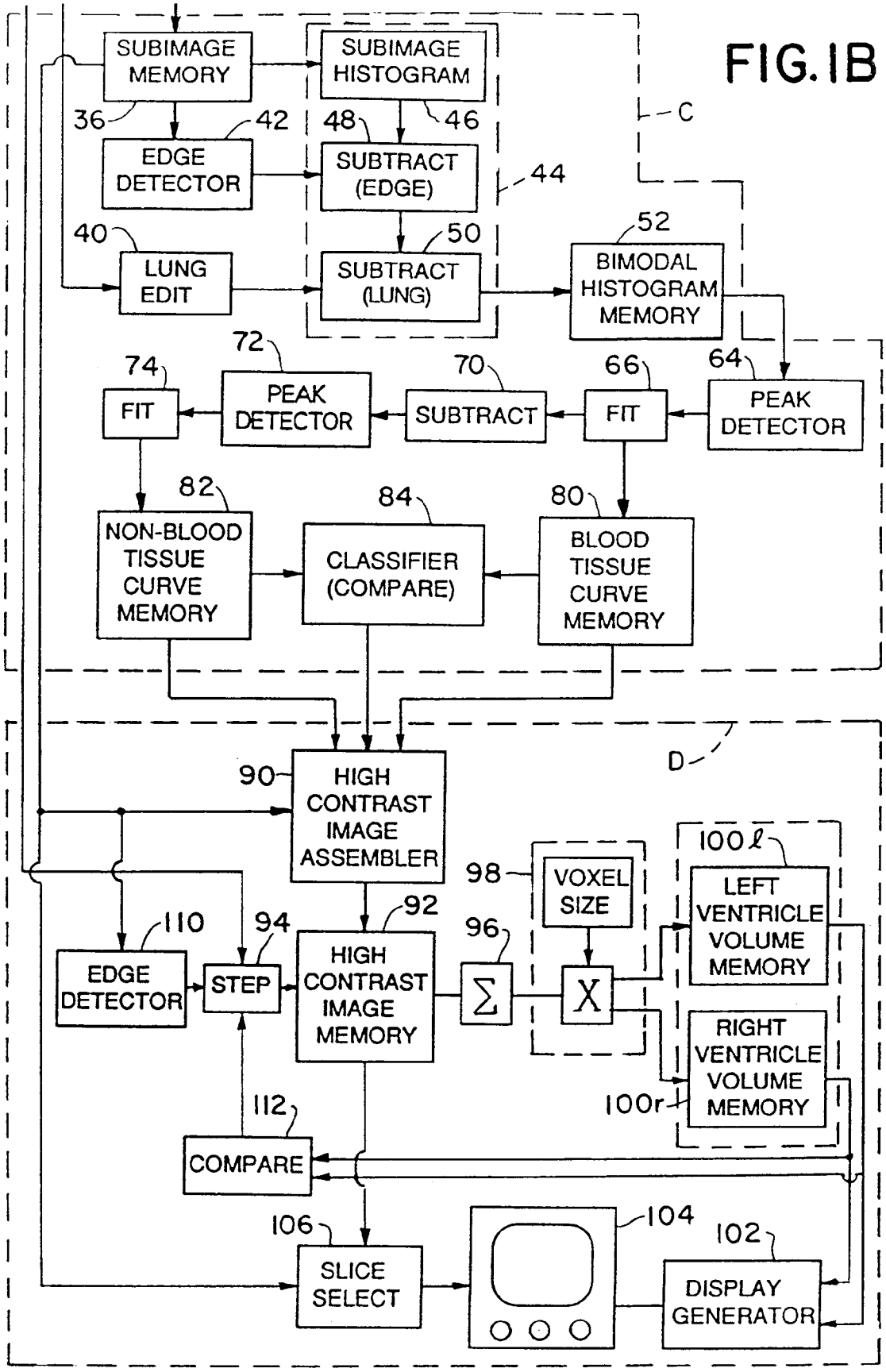
SUBIMAGE MEMORY
SUBIMAGE HISTOGRAM
36
42
48
46
EDGE DETECTOR
SUBTRACT (EDGE)
44
C
40
50
52
LUNG EDIT
SUBTRACT (LUNG)
BIMODAL HISTOGRAM MEMORY
72
74
70
66
64
FIT
PEAK DETECTOR
SUBTRACT
FIT
PEAK DETECTOR
82
84
80
NON-BLOOD TISSUE CURVE MEMORY
CLASSIFIER (COMPARE)
BLOOD TISSUE CURVE MEMORY
D
90
HIGH CONTRAST IMAGE ASSEMBLER
100ℓ
98
VOXEL SIZE
LEFT VENTRICLE VOLUME MEMORY
110
94
92
96
EDGE DETECTOR
STEP
HIGH CONTRAST IMAGE MEMORY
Σ
X
RIGHT VENTRICLE VOLUME MEMORY
100r
112
COMPARE
106
104
102
SLICE SELECT
DISPLAY GENERATOR

FIG. 1B

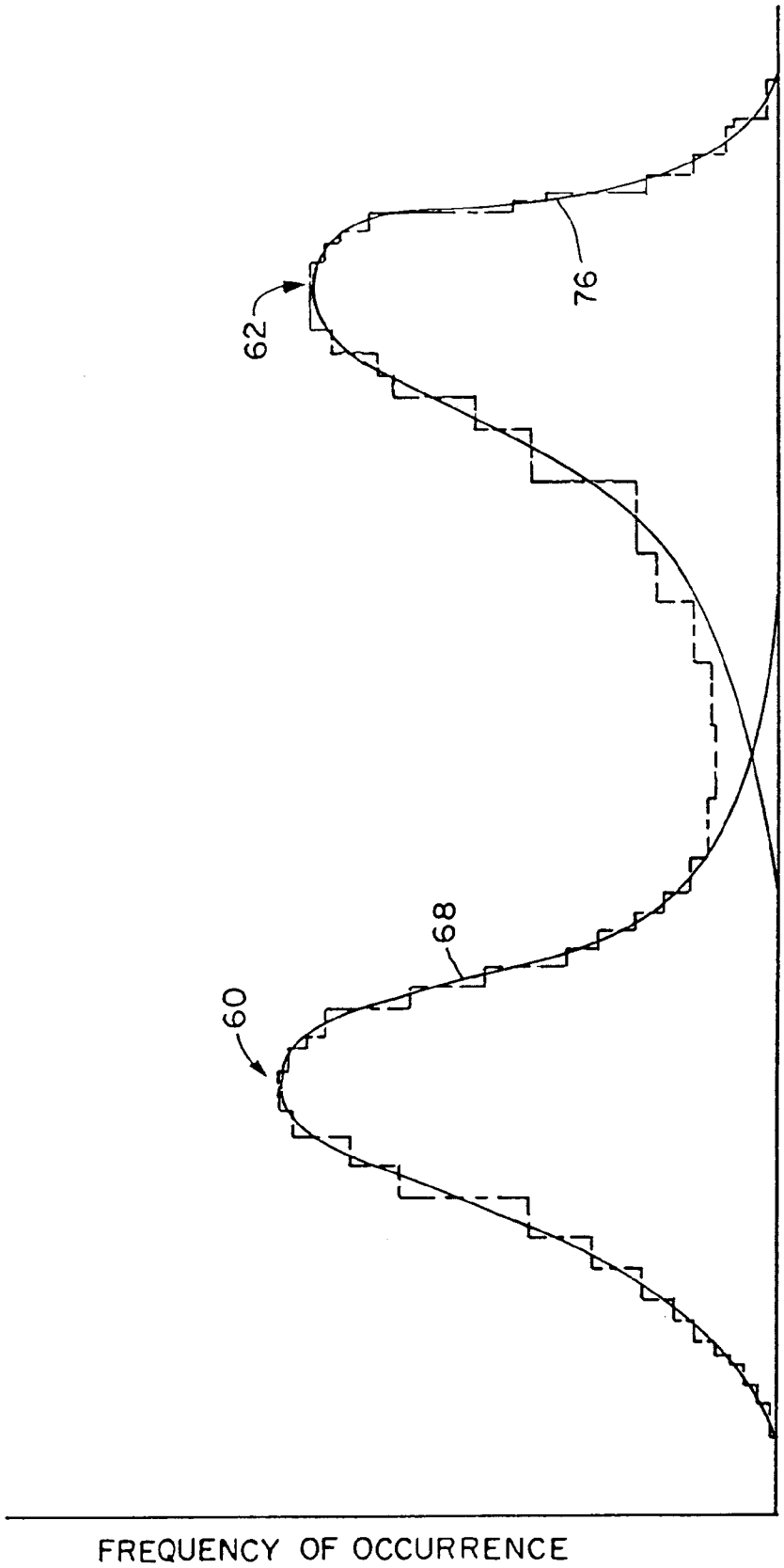

FIG. 2